Effect of Frontal Crash Pulse Variations on Occupant Injuries

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ABSTRACT

The purpose of this investigation was to develop a better understanding of the effect of crash pulse magnitude and shape on occupant injuries. To this end, several idealized frontal crash pulses were used in an occupant simulation, from which the corresponding injury criteria were calculated. The idealized pulses ranged from simple step pulses to two stage pulses that are more comparable to actual vehicle accelerations. Finally, the effect of 5-10 ms duration spikes in different portions of a typical crash pulse was evaluated. From the results of these simulations, several conclusions were drawn.

For the constant acceleration level pulses, the lower magnitude, longer duration pulses resulted in lower injury criteria. However, most crash pulses do not have a single constant acceleration level. For the two stage acceleration pulses, it was found that the injury criteria were reduced as the magnitude of the first stage of the pulse was increased and the level of the second stage was decreased, while holding the total crush space constant. Finally, it was determined that a 5-10 ms spike in the accelerations would significantly affect the injury criteria, regardless of the time at which these spikes occurred.

INTRODUCTION

A better understanding of the effect of the crash pulse shape on the dummy injury criteria can bridge the gap in understanding how changes to a vehicle's structure affect the resulting dummy injury criteria. In pursuit of this understanding, an occupant simulation model was created in TNO's MADYMO™ (version 5.41) [1] multi-body dynamics software code. This occupant model was of a belted driver in a 56 km/h frontal barrier impact. This MADYMO™ occupant model was developed using the same methodology as that used in the development of previous frontal impact occupant models that correlated well with vehicle tests.

Several crash pulses were applied to this occupant model, while holding all other input parameters fixed. These crash pulses varied from simple one and two level step curves to pulses that are more representative of previously measured vehicle frontal crash pulses. The dummy injury criteria calculated in the simulation for these different pulses were compared.

MODEL DESCRIPTION

The MADYMO™ model, pictured in Figure 1, is comprised of a Hybrid III dummy model [1] combined with a typical vehicle interior model, including the occupant restraint system. The vehicle interior model includes the steering wheel, column, floor, knee bolsters, and seat. The occupant restraint system includes the airbag and seatbelt system, which includes a pretensioner and load limiter.

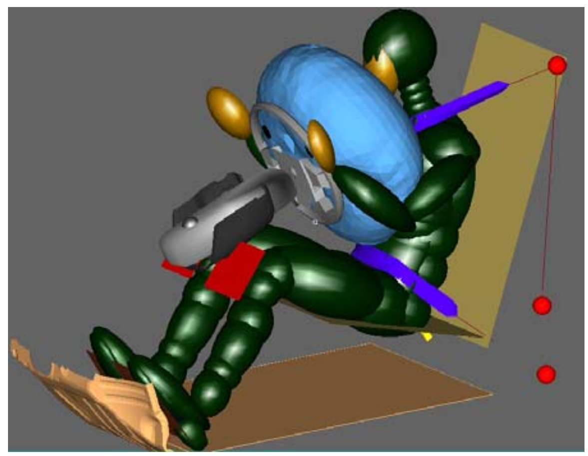

Figure 1. MADYMO™ Model.

The AM50 Hybrid III dummy model used in the simulation was taken from the database provided by the software vendor, TNO [1]. The airbag model is of a typical round driver's side frontal bag with a 200 KPa inflator. The knee bolsters are modeled via planes with force vs. deflection curves. The seatbelts are modeled with finite elements (FE) where they contact the dummy, with MADYMO™ belt model sections connecting the FE portions to the anchor points on the vehicle. A 4KN load limiter was modeled. The deployment times for both the airbag and seatbelt pretensioner are the same for all iterations.

The acceleration pulse is applied to the dummy and vehicle interior, with the vehicle base fixed to ground. The applied crash pulse accelerates the dummy into the vehicle interior, as is typical in MADYMO™ models.

SIMULATION RESULTS & ANALYSIS

Single Step Pulses

The MADYMO™ simulation was initially run with the single step crash pulses shown in Figure 2. Each pulse ramps up and back down from the constant maximum value over a 10-15 ms time period. These rise and fall times were chosen based on passenger car test data, which exhibited similar time durations for transitions between acceleration levels. The area under each of the pulses is the same, matching the area under an actual vehicle crash pulse, and corresponding to the velocity change for this crash event.

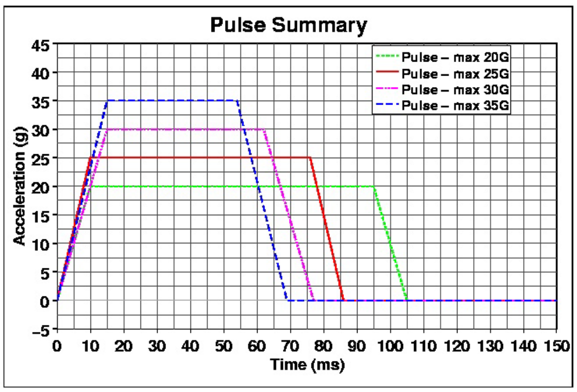

Figure 2. Single Step Pulses.

A comparison of the resulting dummy injury criteria for these single step pulses is shown in Figures 3 and 4. Only the chest and head accelerations are addressed in this paper, both to minimize the number of parameters to be displayed, and because these outputs are representative of the trends. As one might expect, both the chest and head accelerations increase as the magnitude of the pulse is increased.

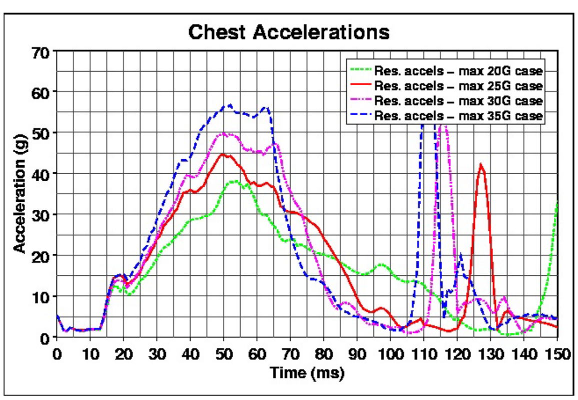

Figure 3. Step Pulse Chest Accelerations.

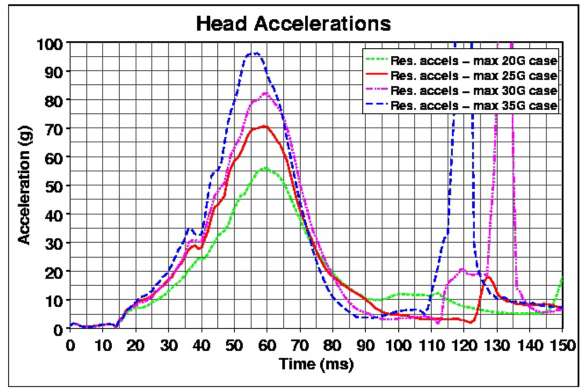

Figure 4. Step Pulse Head Accelerations.

Table 1 summarizes the HIC and 3 ms chest g clip results for these single step pulses. The acceleration spikes occurring after 110 ms in the simulation are due to contact with the steering wheel airbag deflation. These spikes were ignored for the HIC and chest g clip calculations throughout this paper, since they might mask the primary effect of the pulse, and in practice an airbag would be tuned to avoid this situation.

Pulse	HIC	3 ms Chest g Clip
20G Pulse	403.8	37.7 g
25G Pulse	729.7	44.0 g
30G Pulse	1029.8	49.0 g
35G Pulse	1386.2	55.8 g

Table 1. Single Step Pulse HIC and 3 ms Clip.

As Figure 5 shows, there is a clear trend towards lower injury criteria as the magnitude of the applied pulse is reduced. The 20g pulse produces the lowest results. However, it is not always possible to design a vehicle with a constant level crash pulse, or to keep the acceleration level below 20g for this test mode. Therefore, two-step pulses were next analyzed.

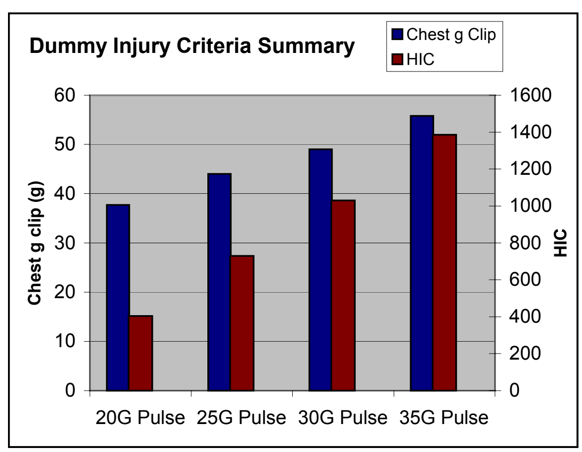

Figure 5. Step Pulse Dummy Injury Criteria.

2-Step Pulses

Most frontal crash pulses are more complicated than a single step pulse, so the next phase in this investigation was to perform simulations with two-step pulses. Figure 6 shows these pulses, along with the 20g single step pulse and a typical passenger car test pulse, for comparison purposes. The two-step pulse shape more closely approximates an actual crash pulse.

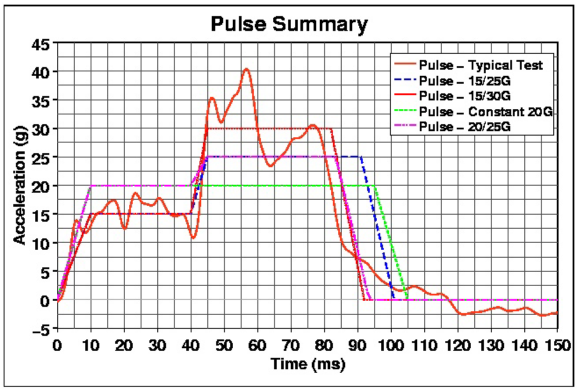

Figure 6. 2-Step Pulses.

The relative magnitudes of the first and second steps of the pulse were varied, and the resulting injury criteria were compared. Simulations were run with 15/25G, 15/30G, and 20/25G pulses. Changes in the step levels were effected over a 5-10 ms time period. The 5 ms transition time from the first step to the second step level was based off the transition time of the test data that is also shown in Figure 6. The duration of the primary step was fixed for all cases, and again based off the test data shown. The assumption behind this was that the longitudinal location of the engine, which drives this timing, would be assumed fixed. The duration of the second step level was then determined such that the required total deceleration would occur.

Figure 8 shows the chest accelerations for the two-step pulses. The 20/25g pulse produces higher peak chest accelerations than the 20g pulse, as one would expect. The 15/25g pulse produces lower chest accelerations than the 20/25g pulse, which is again not surprising since the pulse with the lower combined acceleration levels yields lower chest accelerations. The 15/30g pulse results in higher chest accelerations than those of the 20/25g pulse.

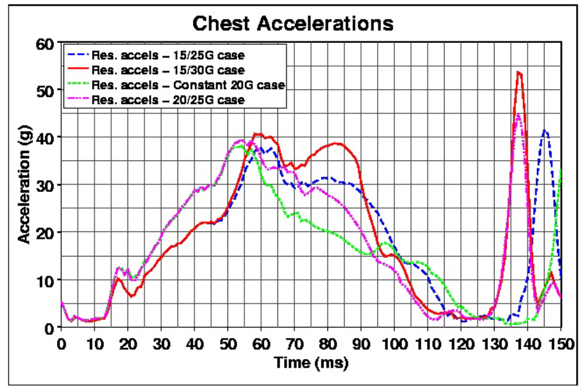

Figure 7. 2-Step Pulse Chest Accelerations.

Figure 8 shows the head accelerations for the two-step pulses. Here both pulses with the higher primary acceleration level produce larger head accelerations. This occurs because the head acceleration peak occurs when the head pitches forward into the airbag, while the dummy's torso is restrained by the seatbelts. Thus the pulses with higher initial primary step levels accelerate the dummy's head to a higher velocity prior to contact with the airbag.

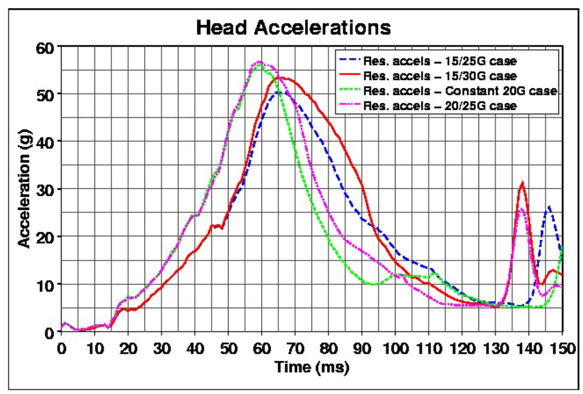

Figure 8. 2-Step Pulse Head Accelerations.

Table 2 summarizes the HIC and 3 ms chest g clip results for these two-step pulses, along with the constant 20g pulse results, while Figure 9 displays these results more graphically.

Pulse	HIC	3 ms Chest g Clip
15/25G Pulse	370.8	36.7 g
15/30G Pulse	492	40 g
20G Max Pulse	403.8	37.7 g
20/25G Pulse	477.5	38.7 g

Table 2. 2-Step Pulse HIC and 3 ms Clip.

All of these pulses produce HIC and chest clip results lower than the constant 25G pulse. This includes the results for the case with a 30g secondary acceleration level.

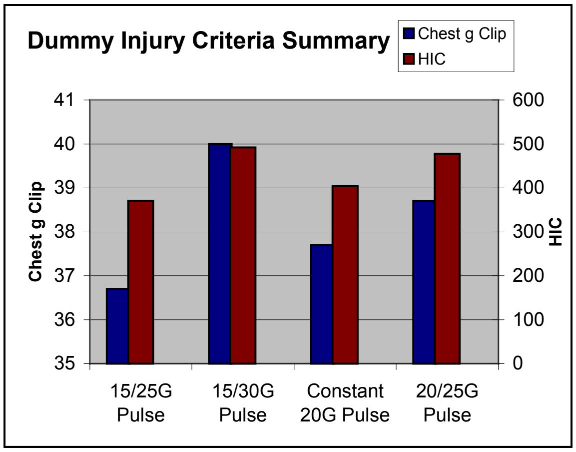

Figure 9. 2-Step Pulse Dummy Injury Criteria.

However, up to this point in this, the crush distance corresponding to these pulses has not been considered. Figure 10 shows a plot of the displacement versus time curves for these 2-step pulses, corresponding to the crush distance of the vehicle. As this plot shows, the crush distances for the different pulses are not equal. Moreover, the 15/25g and 15/30g pulses have larger crush distances than that of the test pulse. Therefore for the next phase of this study, the crush space was restricted to the 712.5 mm of the test data. Thus, the step magnitudes of the pulses were adjusted to reflect this, as described in the next section.

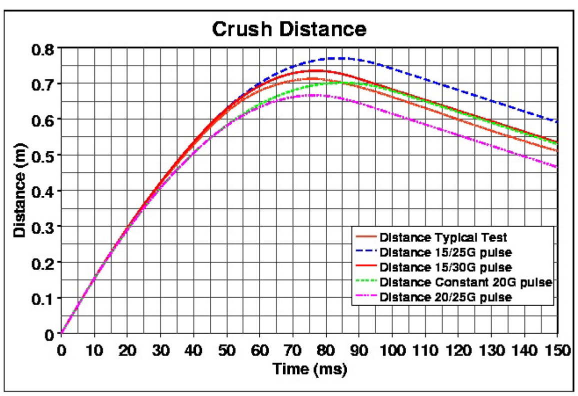

Figure 10. 2-Step Pulse Crush Distance.

Figure 11 shows the velocity traces corresponding to the 2-step pulses, along with that of a typical passenger car test. The 15g initial deceleration level is comparable with the test data.

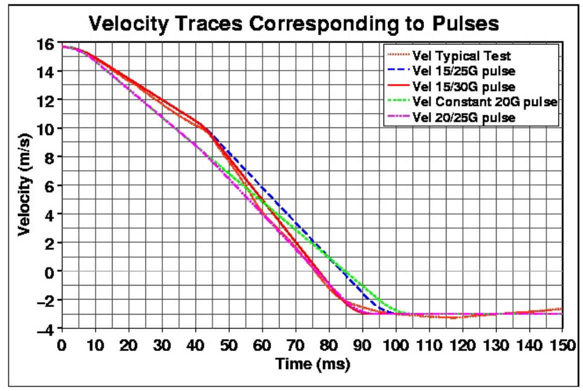

Figure 11. 2-Step Pulse Velocity Traces.

Equal Crush Pulses

Figure 12 shows two-step crash pulses where the magnitudes of the step levels were chosen such that their resulting crush distance was equal to the test data. Figure 13, a plot of the distances, verifies this.

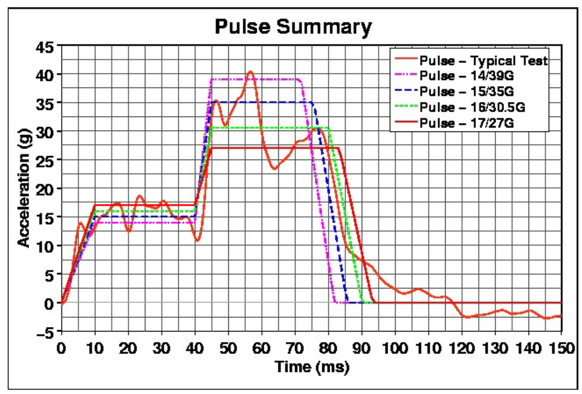

Figure 12. Equal Crush Pulses.

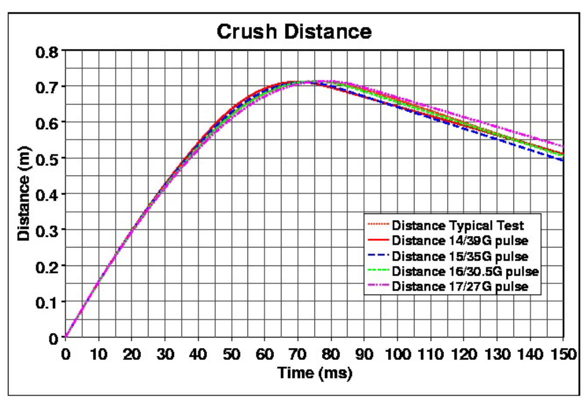

Figure 13. Equal Crush Pulse Crush Distances.

Figures 14 and 15 show the chest and head accelerations for these two-step pulses with the imposed crush distance. Now the trends for both the chest and head accelerations are the same: the peak accelerations are higher for the pulses with lower primary and higher secondary acceleration levels.

Thus, for lower dummy injury values, it is preferable to have a higher initial vehicle deceleration along with a lower secondary deceleration level, while using the same crush space. This might be achieved by increasing the structural stiffness at the front of the vehicle, so that more energy can be dissipated early in the impact event.

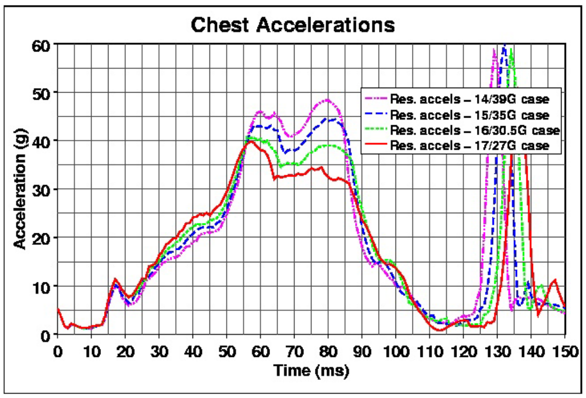

Figure 14. Equal Crush Pulse Chest Accelerations.

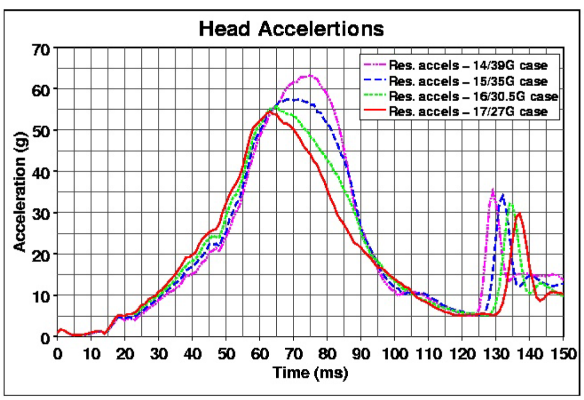

Figure 15. Equal Crush Pulse Head Accelerations.

Table 3 lists the HIC and 3 ms chest g clip for these equal crush two-step pulses, while Figure 16 graphically illustrates the same results. The pulses with the lowest maximum acceleration produce the lowest injury criteria.

Pulse	HIC	3 ms Chest g Clip
14/39G Pulse	696.4	47.9 g
15/35G Pulse	620.4	44.1 g
16/30.5G Pulse	520.6	40.1 g
17/27G Pulse	455.2	39 g

Table 3. Equal Crush Pulse HIC and 3 ms Clip.

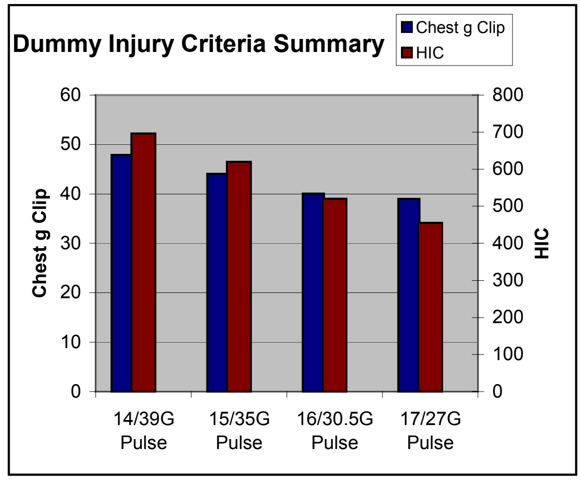

Figure 16. Equal Crush Pulse Dummy Injury Criteria

Pulse Component Analysis

Having established an understanding of the effects of ideal pulses, it is then useful to evaluate how differences between more realistic pulses affect the dummy injury criteria. Two different passenger car crash pulses, A and B, were compared. Pulse A results in simulation dummy injury criteria significantly lower than those of pulse B. Both pulses are shown in Figure 17.

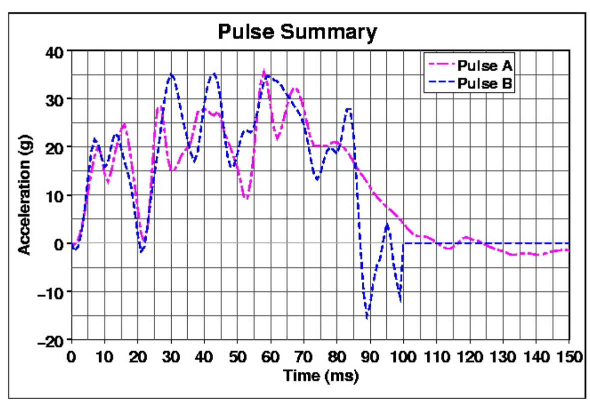

Figure 17. A and B Pulses.

There are two major differences between these pulses. The first difference is the higher magnitude of the acceleration peaks at 30 and 43 ms. The second difference is the longer duration of pulse B's peak occurring at 60 ms, when compared to pulse A, which has two shorter duration peaks over the same time period. The objective is then to determine the individual effect on the injury criteria from each of these differences.

To separately evaluate the contributions to the injury criteria from these differences between the pulses, two new pulses were created. These new hybrid pulses isolate the two different areas of pulses A and B. The first hybrid pulse is pulse A crossing over to pulse B at 45 ms. Similarly, the second hybrid pulse is pulse B crossing over to pulse A at 45 ms. Thus, each hybrid pulse isolates one of the two key areas of pulse B, as shown in Figure 18. Both pulses A and B have nearly the same acceleration level at the 45 ms transition time, which allows for a smooth transition.

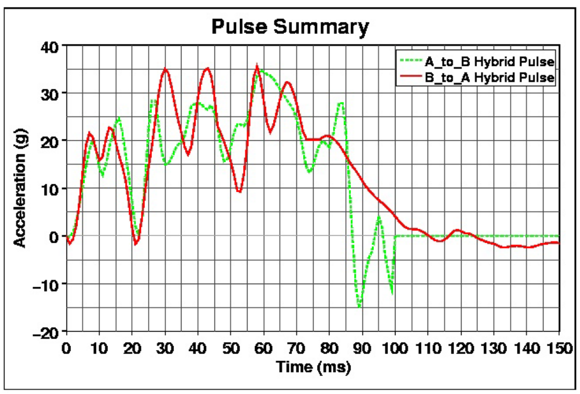

Figure 18. Hybrid Pulses.

Figures 19 and 20 show the chest and head accelerations for the hybrid pulses, along with those for pulses A and B. The peak chest acceleration is higher for both hybrid pulses, although not as high as that of pulse B. The peak head acceleration for either of the hybrid pulses is as high as that of pulse B. However, the hybrid pulse head accelerations are of shorter duration, resulting in a lower HIC value.

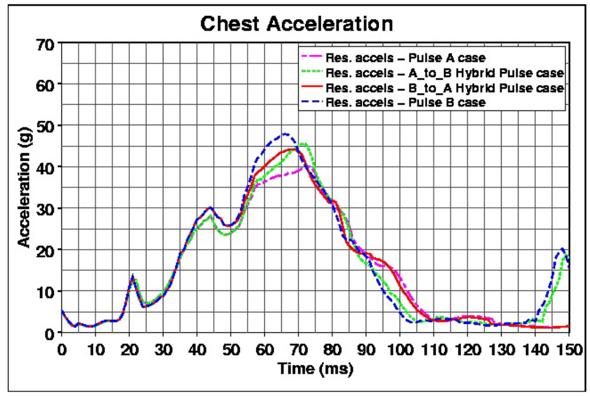

Figure 19. Hybrid Pulse Chest Accelerations.

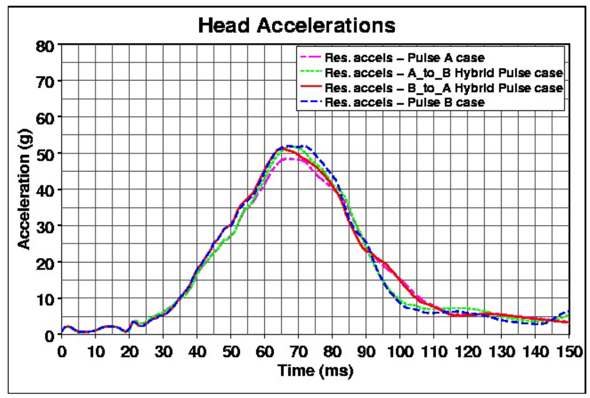

Figure 20. Hybrid Pulse Head Accelerations.

Table 4 summarizes the HIC and 3 ms chest g clip results for pulses A and B and the hybrid pulses, and Figure 21 graphically illustrates these results. Only reducing pulse B's higher earlier peaks (A_to_B pulse) results in 31% of the total reduction of the chest g clip between pulse B and pulse A. Replacing the 15 ms duration pulse with two shorter peaks over the same time span (B_to_A pulse) reduces the chest g clip by 44%. Changing either of these two areas (either hybrid pulse) reduces the HIC by 40.8%.

Pulse	HIC	3 ms Chest g Clip
Pulse A	393	39.6 g
A_to_B Pulse	435	44.9 g
B_to_A Pulse	435	43.9 g
Pulse B	464	47.3 g

Table 4. Hybrid Pulse HIC and Chest g Clip.

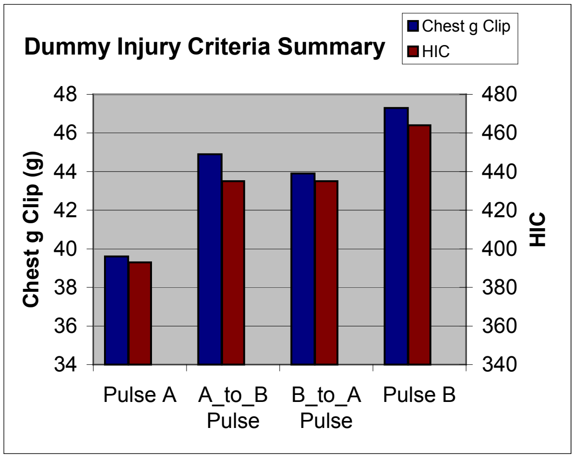

Figure 21. Hybrid Pulse Dummy Injury Criteria.

CONCLUSIONS

This study provides some useful insights into how the shape of the crash pulse affects the dummy injury criteria. For a single constant acceleration level crash pulse, the injury criteria scale almost linearly with the magnitude of this pulse. It was found that that the HIC increased by approximately an additional 85% over the 20G pulse result for each additional 5g increase in the pulse magnitude. Similarly, the chest g clip increased an additional 15% over the 20G pulse result for each additional 5g. However, such a simple crash pulse is not typically seen. This is because the engine normally contacts the barrier wall before the vehicle comes to rest. This engine contact provides an additional load path to the passenger compartment, resulting in at least a second step level in the acceleration pulse.

Given a two-step pulse, the dummy injury criteria may be higher or lower than a constant level pulse, depending on the magnitudes. For a 20/25G pulse, the injury criteria will be higher than those of the 20G constant level pulse; 2.7% for the chest g clip, and 18.3% for the HIC. This represents a 5g increase of the secondary step level. If the primary step level is reduced to 15g, with a 10g increase at engine contact (15/25G pulse), the HIC is reduced 22%, and the chest g clip is reduced 5%. The 15/30G pulse maintains the same primary acceleration level, but assumes a 15g increase at engine contact. This results in a 33% increase in the HIC and a 9% increase in the chest g clip. However, while the 15/25G pulse results in lowest dummy injury criteria for the pulses analyzed, it also has a larger crush distance than the 712.5 mm calculated for the typical crash test pulse used in this investigation.

Given a two-step pulse with a 712.5 mm constraint on the crush distance, the dummy injury criteria are lower for a pulse with a higher initial and lower secondary acceleration level. For example, changing from a 15/35G to a 16/30.5G pulse results in a 16% reduction in the HIC and a 9% reduction in the chest g clip. The higher initial acceleration level reduces the vehicle velocity sooner, leaving less energy to be dissipated during the secondary step level, allowing this portion of the pulse to be lower while remaining within the available crush space. However, the vehicle's structure must still be designed to produce a lower secondary acceleration level along with the higher initial level.

Simulation of the model with pulses obtained from either actual crash tests or computer simulations provided additional insights into how the individual aspects of the crash pulse affect the dummy injury criteria. It was determined that a 7-8g reduction of 5-10 ms duration peaks in the crash pulse will decrease the HIC by 6% and the chest g clip by 5%. It was also found that breaking up a single 15 ms duration acceleration peak into two shorter duration peaks within the same 15 ms time span will decrease the HIC by 6% and the chest g clip by 7%. And simultaneously making both of these changes to the crash pulse will reduce the HIC by 15% and the chest clip by 16%.

Thus an occupant simulation model can be used to evaluate the relative contribution to the injury criteria of specific aspects of the crash pulse. This can help to identify which areas of the crash pulse to try to modify through vehicle structural changes, in order to obtain the largest reduction in dummy injury criteria from such changes.

ACKNOWLEDGEMENTS

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REFERENCES

(1) MADYMO™ Database Manual, Version 5.4, May 1999.